Balanced steady-state free precession spatial gridding

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Introduction: Balanced gradient echo sequences are exquisitely sensitive to external field inhomogeneity (1). This troublesome feature requires first-rate field shimming and hardware to suppress banding artifacts in the acquired image. However, banding can be exploited to create a uniform grid by applying a static field gradient each TR. Here we present a simple modification to the balanced gradient echo which forms image gridlines in the steady-state. Steady-state gridding is potentially useful because the signal is obtained rapidly, without any preparation like RF tagging in SPAMM, with persistent grid lines in the steady-state and high signal-to-noise.

Theory: The bSSFP signal is diminished when the accumulated phase per TR $\Phi_{TR} = \pi$ as shown in Fig. 1. If the gradient moment due to phase and frequency encoding gradients is reset due to balanced gradients each TR, then the total gradient moment is entirely from static field gradient $k_x = \gamma G_x t$. For a static field gradient along a single logical gradient axis, the gradient amplitude G_x and duration t determines the grid spacing $\Delta x = \pi / \gamma G_x t$ and direction x. For three orthogonal, logical gradients, the effective gradient amplitude $G_{eff} = \sqrt{(G_x^2+G_y^2+G_z^2)}$ and duration t determines the grid spacing and direction.

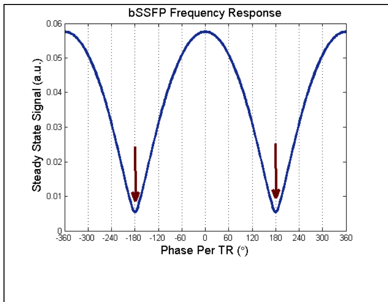


Figure 1: Steady-state signal frequency response of the phase-alternated, balanced gradient echo pulse sequence.

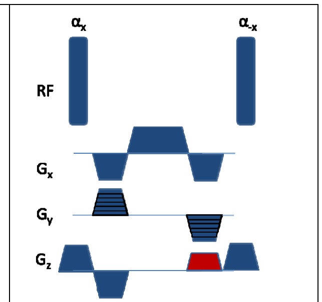
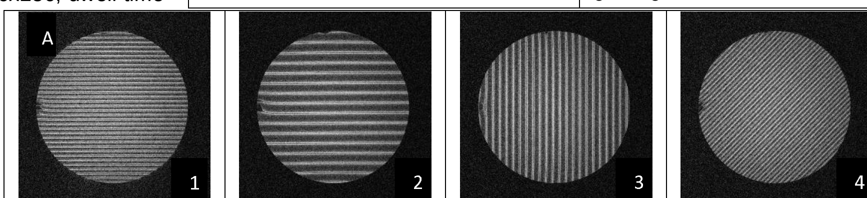
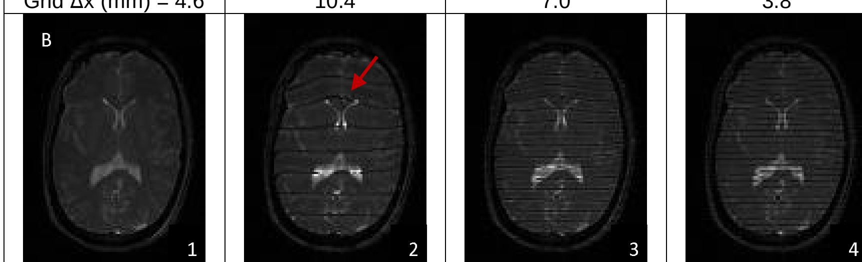


Figure 2: A single TR of the 2D alternating, balanced, steady-state free precession sequence supplemented with an additional static field gradient (red) for gridding.

Methods: Imaging was performed at the University of Pennsylvania on a 1.5 T Siemens Sonata and 3 T Siemens Trio both equipped with nominal 40 µT/mm gradients and either a Siemens body transmit coil and receive head coil (1.5 T) or a Bruker transmit/receive coil (3 T). A doped water phantom (0.05 mM $MnCl^{2-}$, 150 mM ^{23}Na) was used for all phantom experiments at 1.5 T. Four healthy subjects gave informed consent prior to the 3T MRI scan. The phase-alternated balanced gradient echo sequence parameters were TE/TR = 6/12 ms, flip = 20°, slice thickness = 8 mm, FOV = 200 mm^2, matrix = 256x256, dwell time = 10 µs. The sequence was prepared with a half-alpha RF pulse and 256 dummy RF pulses. Each TR an additional field gradient was pulsed immediately after frequency encoding as shown in Fig. 2.



Duration (ms) = 0.5	0.45	0.45	0.45
Grad (µT/mm) = 10	5	7.5	G_x = 10; G_y = 10;
Grid Δx (mm) = 4.6	10.4	7.0	3.8



Time (ms) = 0.45	0.45	0.45	0.45
Grad (µT/mm) = 0	5	10	15
Grid Δx (mm) = NA	10.4	5.2	3.4

Results: Fig. 3a and Fig. 3b illustrate balanced gradient echo gridding in both an agarose phantom (1.5 T) and an *in vivo* human brain (3 T). For both objects, the grid spacing Δx is proportional to the gradient pulse duration and amplitude. Gridlines form along frequency encoding x (Fig. 3A-1, Fig. 3A-2), phase encoding y (Fig. 3A-3), combination (Fig. 3A-4) or in oblique directions in 2 or 3 dimensions (not shown) or combine gridlines along multiple directions by combining kspace data from multiple acquisitions (not shown). With a small accumulated gradient moment, the effects of external field inhomogeneity manifested as grid curvature (Fig. 3B-2 - red arrow), but the effects are reduce as the effective field is increased (Fig. 3B-3,4). Grid spacing also depends on the appearance of the steady-state frequency response, which may not be shaped like Fig.1.

Discussion: Steady-state gridding includes all the benefits of balanced gradient echo acquisitions, particularly rapid acquisition times (< 2 minutes for a full 3D acquisition), high signal to noise, and persistent grids in the steady-state with a decrease in conventional T1/T2 contrast. In future work we will explore the grid sensitivity to motional processes such as translation, rotation in cardiac and lung motion and flow. In the limit that the gridding gradient is large, the unbalanced gradient moment effectively serves as a spoiler and the signal adopts the SSFP-fid response. This is equivalent to the condition that the grid line spacing becomes smaller than the voxel size. This condition has been shown to reduce bSSFP signal banding due to static field inhomogeneity (2).**References:** (1) Haacke, et al. MRI: Physical Principles and Sequence Design. (2) Zur, Y, et al. Magn. Res Med. 6, 175-193 (1988).